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Original Article

The effects of an eight-week strength training intervention on tibiofemoral joint loading during landing: a cohort study

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Abstract

Objectives: To use a musculoskeletal model of the lower limb to evaluate the effect of a strength training intervention on the muscle and joint contact forces experienced by untrained women during landing.

Methods: Sixteen untrained women between 18 and 28 years participated in this cohort study, split equally between intervention and control groups. The intervention group trained for eight weeks targeting improvements in posterior leg strength. The mechanics of bi- and uni-lateral drop-landings from a 30 cm platform were recorded pre and post intervention, as was the isometric strength of the lower limb during a hip extension test. The internal muscle and joint contact forces were calculated using FreeBody, a musculoskeletal model.

Results: The strength of the intervention group increased by an average of 35% ($p < 0.05$; pre: 133±36 N, post: 180±39 N), whereas the control group showed no change (pre: 152±36 N, post: 157±46 N). There were only small changes from pre to post test in the kinematics and ground reaction forces during landing that were not statistically significant. Both groups exhibited a post test increase in gluteal muscle force during landing, and a lateral to medial shift in tibiofemoral joint loading in both landings. However, the magnitude of the increase in gluteal force and lateral to medial shift was significantly greater in the intervention group.

Conclusion: Strength training can promote a lateral to medial shift in tibiofemoral force (mediated by an increase in gluteal force) that is consistent with a reduction in valgus

loading. This in turn could help prevent injuries that are due to abnormal knee loading such as anterior cruciate ligament ruptures, patella dislocation and patellofemoral pain.

Summary Box

- Strength training of the lower limb resulted in a lateral to medial shift of tibiofemoral forces during drop-landing.
- This appeared to be mediated by an increased force in the gluteal musculature during landing.
- Musculoskeletal modelling of the lower limb can demonstrate changes in lower limb mechanics during drop-landing that have not been reported using traditional methods.

Introduction

Abnormal knee joint loading has been shown to be a mechanism of injury in a range of complaints including anterior cruciate ligament (ACL) rupture, patella dislocation and patellofemoral pain [1–4]. Consequently, there has been great interest in finding ways to modify internal joint loading in order to prevent these injuries. However, the outcome measures of such studies have generally been the calculation of external kinematics and kinetics or inter-segmental mechanics (i.e. joint angles, inter-segmental forces and moments calculated by inverse dynamics analysis, or ground reaction forces; GRF [5–7]). Although useful, these calculations do not indicate the actual loading experienced by the internal structures of the knee (i.e. the forces experienced by muscle-tendon units, ligaments and bones). For instance, ACL injury prevention programmes have been shown to successfully modify kinematic outcomes towards movement strategies of lower risk [7,8] and there is epidemiological evidence that such interventions effectively reduce the ACL injury rate [9–11] however, the effect of such programmes on the actual internal joint loading is largely unknown.

Muscle strength and activation are variables that can be directly changed by training programmes [12], and can provide protection against injury in activities like landing from a jump. For instance, previous ACL injury research has described the importance of gluteal and hamstring strength [13,14] and increased hamstring activation pre- and post-landing [15] in reducing injury. Similarly, gluteal activation and strength have been related to a reduction of knee valgus [16], patellofemoral pain [17,18] and patellar dislocation [19] in various activities. Despite these positive associations however, the literature relating to the effect of strength training alone on kinematics and GRF during movement is equivocal [20,21] and the effect on internal knee joint forces is again unknown. To this end, this study employed a posterior lower limb focussed training intervention which would be expected to increase the strength of the gluteal and hamstring musculature.

One technique that can be utilised to estimate internal forces is musculoskeletal modelling and musculoskeletal modellers envisage a future where their work can inform clinical practice [22,23]. For instance, there have been a number of studies that have sought to quantify the forces present in the knee during landing [24–29]. However, no study has used musculoskeletal modelling technology to assess the effect of a posterior thigh musculature focused training intervention on the forces experienced by the internal structures of the knee. The objective of this study was therefore to evaluate the effects of a leg strength training intervention on internal knee forces during landing (tibiofemoral joint reaction forces; TF) using a publicly available musculoskeletal model of the lower limb [30]. We hypothesized that the intervention would result in a lateral to medial shift in TF that is consistent with the changes in landing mechanics that have previously been seen after strength training [21,31].

Methods

Experimental approach

This study was divided into three phases undertaken at St Mary's University. Firstly, during the pre test the performance of the participants in a landing task was assessed alongside a measure of their posterior lower limb strength. Next, the experimental group took part in an eight-week training intervention designed to increase their posterior lower limb strength whereas the control group kept up with their usual recreational activities. Finally, all participants were retested using the same protocol as in the pre test. The experimenters were not blinded as to the participant groups.

Participants

Sixteen young, healthy students participated in this study (Table 1) and were assigned to either the control group (CG) or intervention group (IG) based upon their availability to take part in the intervention training programme. The recruitment criteria stipulated that the participants were female, between 18 and 28 years of age, free from musculoskeletal injuries over the preceding 6 months, right foot dominant, and only took part in recreational physical activity (i.e. no heavy resistance or injury prevention training for at least 6 months prior to the study, and that they participated in mainly leisure sports at most four times per week). All participants provided informed written consent prior to the experiment and the ethics sub-committee of St Mary's University approved the study.

Table 1. Participant characteristics (mean ± standard deviation). There were no significant differences between groups ($p > 0.05$).

	Age (years)	Body mass (kg)	Height (m)
Control group	22.9 ± 2.4	62.2 ± 8.3	1.66 ± 0.07
Intervention group	22.0 ± 3.2	65.4 ± 7.1	1.68 ± 0.03

Instrumentation

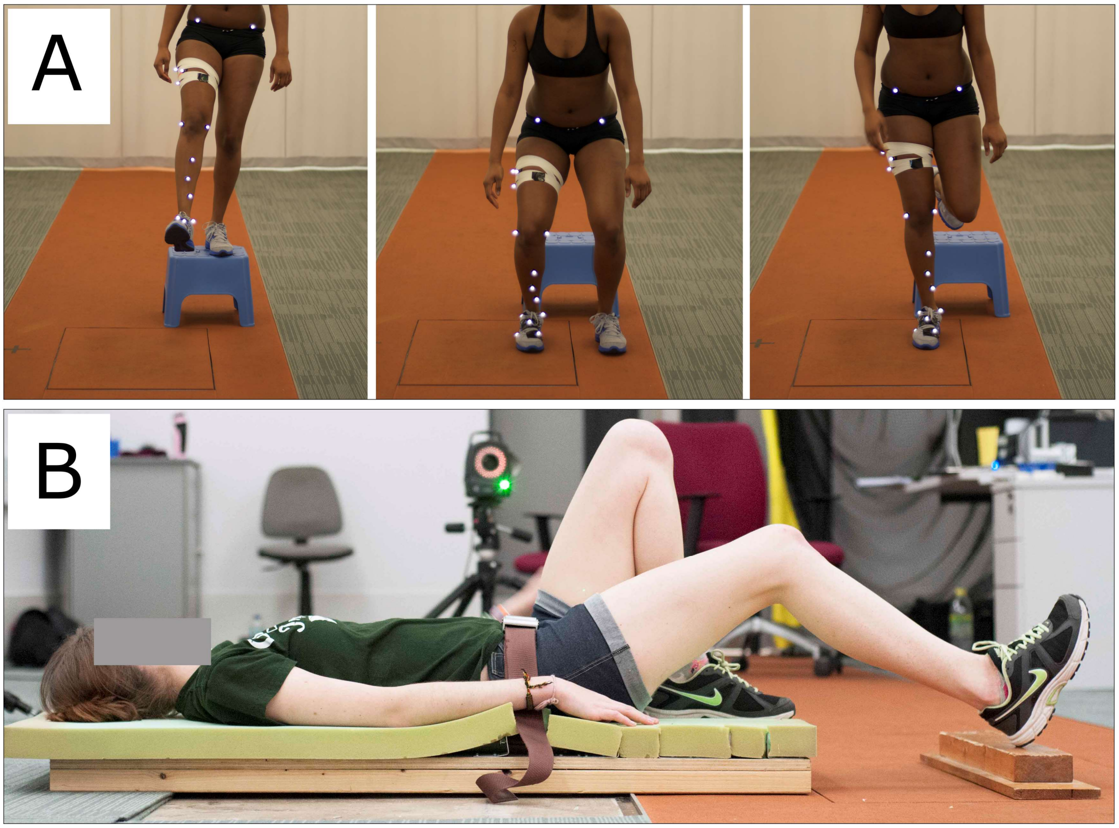
Evaluation of drop landing performance: The kinematics describing the time history of the position of 18 reflective markers (14 mm) placed on key anatomical landmarks of the right leg and pelvis [30] according to the guidelines of Van Sint Jan [32,33] was obtained using a Vicon 3D motion analysis system (Vicon MX System, Vicon Motion Systems Ltd, UK) incorporating 11 cameras. The GRFs during landing were measured with a force plate (Kistler 9287BA Plate, Kistler Instruments Ltd., UK) synchronized with the Vicon system. All data was collected at 200 Hz.

Lower limb strength testing: The strength of the posterior aspect of the lower limb was tested in a closed kinetic chain task as described below using the same Kistler force plate as for the evaluation of the drop landings.

Procedures

After performing a 10-minute supervised, dynamic warm up including running, high knees, buttock kicks, lunges, squats, straight leg walks and hop and stick, the participants practiced the drop landings for up to five attempts both bi- and unilaterally. A three to five minute rest followed, in which the reflective markers were placed on the anatomical landmarks with double-sided adhesive tape. Drop landing data was collected during controlled falls from a 30 cm platform placed 0.5 cm in front of the force plate. Participants first completed five bilateral landings (BLs) and then five unilateral landings (ULs) having been instructed to step forward from the platform with their dominant right foot (and not to jump forwards or step down), land naturally with only their dominant foot touching the force plate and stay in this landing position for at least 2 seconds. During BLs, the participants were asked to land with both feet at the same time (Figure 1A – note the position of the feet with just the dominant foot on the force plate). Incorrect landings contrary to the description above were repeated. The rest periods between the five drop landings for each condition were at least 60 seconds long, and at least two minutes rest was taken between the BLs and ULs.

Figure 1. Experimental arrangements: A. Bi- and unilateral drop landing tasks; B. Assessment of posterior thigh strength utilising a hip extension test.



After a three to five minute rest period, the strength of the right posterior thigh was assessed in a hip extension test. The hip was positioned at a flexion angle of 30° (note in this article we use the convention that when the subject is stood in the anatomical position their ankle, knee and hip joint angles are 0°, and that flexion of the joint is represented by a positive angle). The ankle was positioned neutrally (i.e. at a flexion angle of 0°) with the heel at the centre of a wooden block that was on top of the force plate (Figure 1B). The participants were then encouraged to push the heel downwards with maximum force for a period of at least six seconds and the peak force was recorded. A two minute rest period was taken between the three trials. This hip extension test was chosen as it has previously been shown to be reliable [34] and tests the strength of the limb in a closed kinetic chain task at similar joint angles to those found at initial contact during BL in females [35,36].

Exercise intervention: Eight participants performed an eight-week posterior leg strength programme (Table 2), attending three hourly sessions per week that were supervised by a UK Strength and Conditioning Association qualified coach. Loading was progressed weekly by increasing the load lifted based on individual responses to training (strength, experience and motivation), but sets, reps, rest and perceived exertion were similar within the group.

Table 2. The strength training programme followed by participants in the intervention group.

Week 1-4	Week 5-8	Sets	Reps	Rest
Session 1				
Split Squat	Lunge	3	10	2 min
Good Morning	Ecc/con leg pull&push in pairs	3	10	2 min
SL SLDL	Bulgarian Split Squat	3	10	2 min
Session 2				
Step up (L to M height plyometric box)	Step up (M to H height plyometric box)	3	10	2 min
Nordic hamstring (ecc+con)	Nordic hamstring (ecc+con)	3	6/8	2 min
SL Bridge	SL Good Morning	3	10	2 min
Session 3				
Squats	Squats	3	10	2 min
SLDL	SLDL	3	10	2 min
SL Good Morning	SL Hip thrust	3	10	2 min

SL= single leg, SLDL= stiff leg deadlift, ecc= eccentric, con= concentric, L= low, M= medium, H= high

Data analysis

Musculoskeletal model: In order to compare predicted muscle and joint reaction forces pre and post intervention, the data collected was analysed using a publicly available musculoskeletal model of the lower limb [30,37–40] (FreeBody; www.msksoftware.org.uk). The validation and verification of FreeBody has been described previously [41–44], with a focus on the accuracy of the TF predictions [41] and the sensitivity of the model to the input kinematic data and its muscle force upper bounds [43].

FreeBody represents the lower limb as a linked chain of five rigid segments. The position and orientation of the pelvis, thigh, calf and foot segments at each moment in time are determined from the marker data (the position of each segment has 3 degrees of freedom and its orientation has a further 3 degrees of freedom). The position and orientation of the patella segment is determined based upon the knee flexion angle [30], using relationships developed from previous literature [45,46]. The anthropometry of each segment is determined from the work of de Leva [47]. Given the time history of the position and orientation of each segment and its anthropometry, the kinematics of each segment is calculated using the method of Dumas and colleagues [48]. Next, the data of Klein Horsman and colleagues [49] is used to determine the origins, insertions and lines of actions of 163 muscle elements and 14 ligaments.

Following the above steps the equations of motion governing the movement of the segments can be determined (Equation 1; Appendix). However, there are more unknown forces (193) than there are equations (22), and thus this is an indeterminate problem with many possible solutions. The next step is therefore to pick the most physiologically likely solution. Firstly, the potential solution set is narrowed by imposing physiologically based constraints then the most physiologically likely solution is determined by using an optimization procedure developed [37] from the work of Crowninshield and Brand [50] and Raikova [51] that is implemented using MATLAB (R2013a, Mathworks, 1 Apple Hill Drive, Natick, MA 01760, US). The optimization is predicated upon finding the solution that minimises a cost function based upon maximising muscular endurance (Equation 2; Appendix).

Data processing: For each subject, each landing (BL, UL) and both pre and post tests, the trial that resulted in the lowest peak GRF was selected for analysis (as this was taken to be the most successful landing). A 4th order dual low pass Butterworth filter with a cut off frequency of 6 Hz was used to filter the kinematic and kinetic data. The filtered data was then processed through FreeBody. The strength capabilities of FreeBody (as represented by the maximum force that each muscle and ligament was permitted to experience) were scaled to reflect the participants' strength testing results). Following the example of our previous work, if the optimization routine employed by FreeBody (fmincon routine in MATLAB) could not find a feasible solution for a particular frame then we raised the strength upper bound for the frame until a solution could be found. This was only necessary for a limited number of frames.

Statistical Analysis

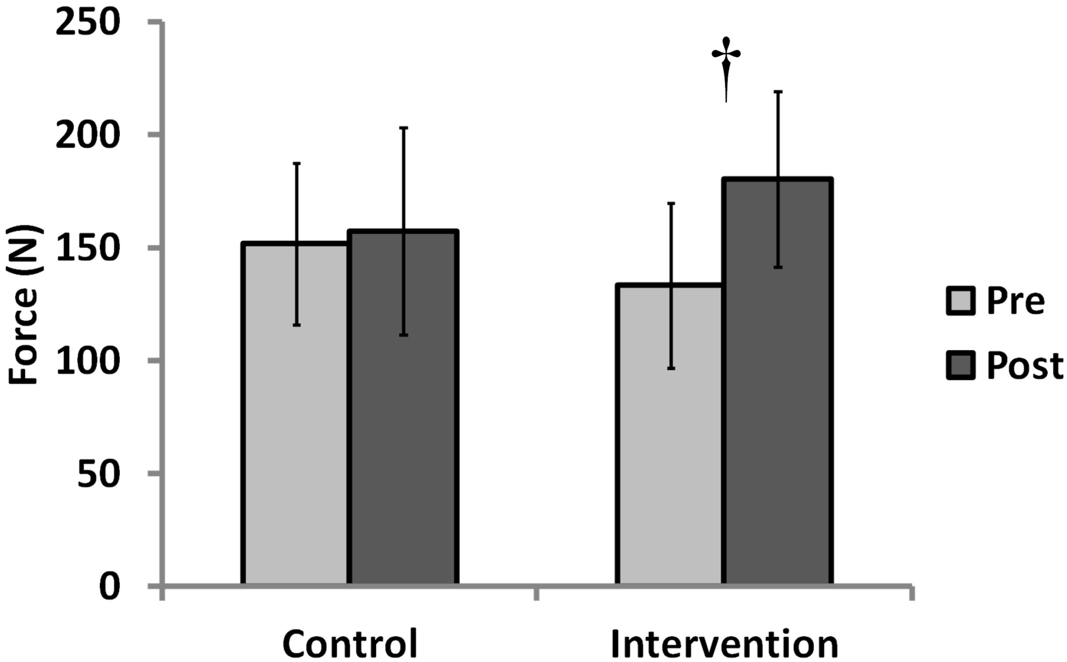
Statistical analysis was performed using IBM SPSS Statistics (version 22, International Business Machines Corp., New Orchard Road, Armonk, NY 10504, US) and MATLAB (R2013a, Mathworks, 1 Apple Hill Drive, Natick, MA 01760, US). ANOVA was used to check for differences in age or anthropometry between the groups at pre-test. An ANCOVA was used to evaluate the change in strength of the right posterior thigh musculature where baseline strength was included as a covariate. The alpha level was set at $p < 0.05$ *a priori* and normality was confirmed by Shapiro-Wilk tests.

The output data from the musculoskeletal model was first normalised with regards to time. A cubic spline was then fitted to each data series and used to interpolate the normalised curves to obtain values at regular intervals. The mean and the 95% confidence interval (CI) at each time point was then calculated for each data series. A significant difference between curves was determined when there was no overlap between the confidence intervals.

Results

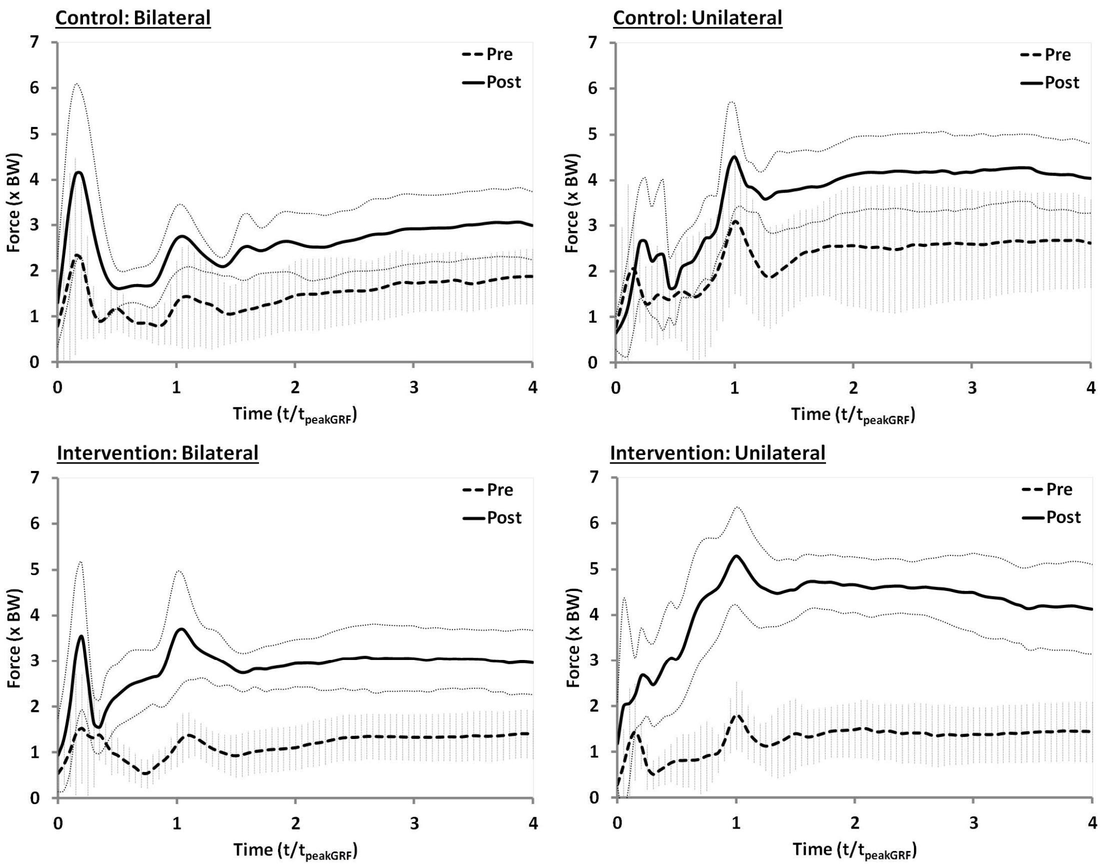
During the intervention the strength of the IG increased by 35% ($p = 0.001$; pre: 133±36 N, post: 180±39 N). There was no change in the strength of the CG (pre: 152±36 N, post: 157±46 N). The participants attended 94% of the planned sessions.

Figure 2. Strength testing results (error bars indicate the standard deviation). † indicates a significant difference between the pre and post test scores of the intervention group (p = 0.001).



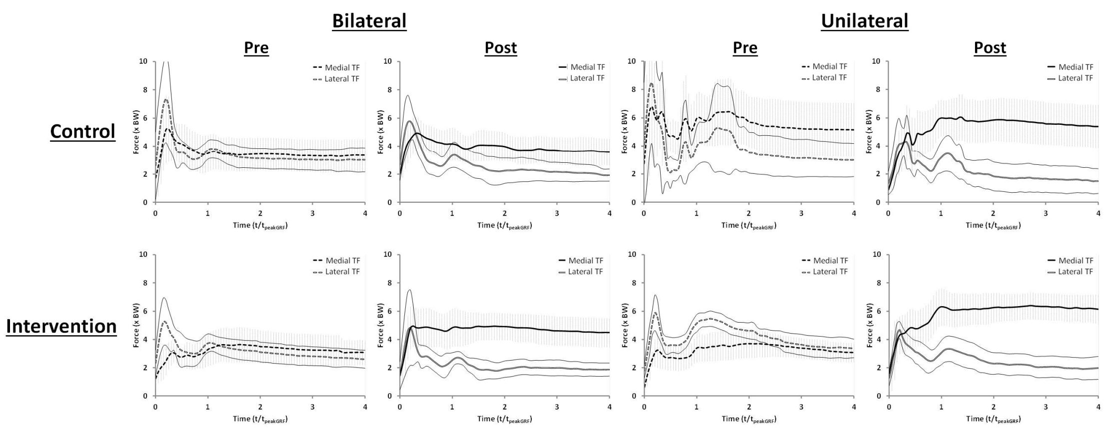
Both CG and IG exhibited an increased use of the gluteal musculature from pre to post test (Figure 3). However, the magnitude of the increase was greater for the IG in both BLs and ULs, and there was also little overlap of CIs (whereas for the CG it was considerable). There were no other strong trends in terms of changes in muscle forces from pre to post test (Web Supplementary Material).

Figure 3. Force in the gluteal musculature during bilateral and unilateral landings. The vertical error bars represent the 95% CI for the pre test, whereas the light dotted lines represent the 95% CI for the post test.



During the pre test, the peak lateral tibiofemoral joint contact force (lateral TF) was greater than the peak medial tibiofemoral joint contact force (medial TF) for all groups (Figure 4). For the CG, the lateral TF then dropped below the medial TF after the first local peak in GRF during both landings. For the IG BL, the lateral TF dropped below the medial TF after the second local peak in GRF, whereas for the IG UL, the lateral TF was greater than the medial TF throughout the analysed time period. During the post test, the lateral TF fell relative to the medial TF for all groups, however the magnitude of this change was greater for the IG than the CG, and greater for the UL than the BL. For the IG, the lateral TF was equal to or lower than the medial TF throughout the time period for both landings.

Figure 4. Lateral and medial tibiofemoral joint reaction forces during bilateral and unilateral landings. The vertical error bars represent the 95% CI for the medial tibiofemoral force, whereas the light dotted lines represent the 95% CI for the lateral tibiofemoral force.



There were only minor differences between the pre and post intervention GRFs for both landing styles and groups (Web Supplementary Material). There was a trend towards slightly higher peak GRFs post intervention during the BLs for both groups (approximately 0.3-0.4 × body weight; BW). In addition, the GRF for the CG UL was marginally lower during the post test (around 0.2-0.3 × BW for most of the time during the landing period). This study was largely unable to demonstrate changes in kinematics between the pre and post test, although both groups showed a trend towards lower hip and knee flexion during BL (Web Supplementary Material).

Discussion

This study supports the hypothesis that TF patterns would be altered following a strength intervention and that these changes would be consistent with the kinetic and kinematic changes that have been previously found to occur after strength training. In particular, we found changes in gluteal muscle forces, and a lateral to medial shift in TF. In contrast, there were only small changes in GRF and the kinematics of landing.

A lateral to medial shift in tibiofemoral joint loading

The most novel result in this study is the change in the pattern of TF after the intervention. Both groups experienced a reduced lateral TF during the post test, however the decrease was greater in the IG than in the CG. In addition, the IG experienced an increase in the medial TF at post test, whereas the medial TF remained similar for the CG. Taken together, these data indicate a lateral to medial shift in knee loading which was of significantly greater magnitude in the IG. Such a shift is consistent with a reduction in knee valgus, although we were unable to detect differences in kinematics. Both groups also experienced an increase in gluteal force post intervention and it has been suggested that increased gluteal force can reduce valgus loading of the knee. The changes in both groups may be explained by a learning effect of the tasks in the post test, however, the fact that the IG experienced greater changes in gluteal force and lateral to medial shift suggests that there was an effect of the intervention. The results of the present work tend to support the link between gluteal force and the medial/lateral loading distribution of the tibiofemoral joint. In addition, these results suggest that strength training can facilitate women in using the gluteal musculature during landing in a way that possibly exhibits a lower risk of knee joint injuries such as ACL rupture, patella dislocation and patellofemoral pain.

The fact that a lateral to medial shift in knee loading was found when there was an increased gluteal force (in both groups) is remarkably consistent with contemporary thinking. For instance, studies have identified relationships between increased hip strength/activation and improved neuromuscular alignment and control of the legs [17] and increased gluteus medius activation and decreased TF [52]. These studies in combination with our results suggest that a stronger posterior hip musculature can result in greater gluteal force expression, altered lateral to medial TF distribution and potentially affect valgus loading.

Effect of strength training on landing kinematics and GRF

There were only small differences in landing kinematics pre to post intervention in both groups (frontal, sagittal and transverse plane), which is similar to another study that could not demonstrate knee valgus/varus and knee/hip extension/flexion changes following a strength training programme [20]. In contrast, one other study did show kinematic alterations of increased hip flexion at initial contact, and peak hip and knee flexion after a basic strength training programme [21] (it should be noted that the programme employed in that study also included flexibility and balance training). The majority of prevention studies that found consistent alterations in kinematics included neuromuscular and feedback training which were not employed in our study [7,53,54]. The lack of kinematic differences in this study, despite the changes of internal kinetics, are important and suggest that either strength training in isolation does not affect kinematics, that kinematics are less sensitive to strength changes than internal kinetics or that musculoskeletal models of the type employed here are more sensitive to changes in internal kinetics than kinematics.

As described above, the inability of this study to demonstrate statistically significant differences in knee varus/valgus is consistent with previous studies that have looked at the effect of strength training [20,21]. One reason for this may be the fact that optical motion capture methodologies are less able to discriminate between differences in internal/external rotation and ab/adduction than between differences in joint flexion and extension due to the measurement error associated with soft tissue artefact [55]. In contrast, we have previously shown that the forces predicted by the model employed here are sensitive to small changes in kinematics (in particular, that they are sensitive to small changes in the internal/external rotation of the tibia [43]). It is thus entirely credible to suggest that musculoskeletal models may be more sensitive to changes in internal kinetics than more traditional approaches are to changes in kinematics. This may have important consequences for future assessment methods, particularly if ACL and knee injury risks are only assessed through a consideration of kinematic factors; in particular suggesting that clinical assessment methods should also incorporate the prediction of internal joint kinetics. The greater sensitivity could be used as an early indicator to prevent knee injuries and may detect smaller changes following intervention programmes. Consequently, this new perspective on joint conditions may offer greater detail in clinical diagnoses.

We were also unable to identify changes in GRF patterns pre and post intervention - this is in agreement with results of other studies that studied limb strengthening interventions [20,21], although contrary to a study that also focussed on posterior thigh musculature [56]. Our findings suggest that either the change in force distribution between the joints altered due to internal modifications as GRF patterns stayed relatively constant or that the internal forces are particularly sensitive to small changes in GRF. Studies that found changes in GRF mostly included feedback or plyometric training, that probably included landing feedback training [53,54,57]. This might suggest the necessity to incorporate direct feedback of landing technique if substantive changes in ground force application are a goal for the patient or athlete.

Role of musculoskeletal modelling in clinical research

As far as we are aware, this is the first study that has used musculoskeletal modelling technology to assess the results of an exercise intervention. The unique finding of this study is the change in lateral to medial loading of the tibiofemoral joint following strength training. This is an observation that is previously unreported, probably due to the fact that other similar studies have relied upon kinematic measurements. Similarly, we have recently successfully employed the same musculoskeletal model as in this study to report the effects of an acute intervention on muscular forces during explosive activity [58]. Taken together, these studies therefore demonstrate the unique sensitivity and potential for musculoskeletal models to improve the understanding of problems with clinical relevance. However, to date we have only used this model to study differences at the cohort level. The employed model incorporates limited subject-specific detail, and thus is currently unable to be used at a subject-specific level. Future work should establish the detail that is necessary to produce such specified results.

Conclusions

In summary, this study demonstrates that a training intervention with a focus on posterior thigh strength resulted in a greater estimated use of the gluteal musculature during drop landings. This was commensurate with an altered pattern of joint loading; in particular, there was a change in force distribution at the tibiofemoral joint with a shift from lateral TF to medial TF, a change that is consistent with a reduced valgus and an increased hip joint loading. Potentially, this could reduce abnormal knee loading injuries that are related to valgus/varus forces such as ligament injuries (i.e. ACL), kneecap dislocation, menisci and cartilage damage. To our knowledge, this is the first time a change in the medial/lateral loading of the knee has been observed following a period of strength training. It is noteworthy that the changes in the internal force loading of the lower limbs were found despite there being only small concurrent changes in GRF and kinematics. This suggests that the joint loading may be more sensitive to changes in strength than kinematic measures, and that clinicians should be mindful when relying solely on kinematic measures.

Competing interests

The authors declare that they have no competing interests.

Contributorship

MC, JG and DC conceived of and designed the study. JG and AB created and validated the strength test used in the study. DC and AB created and tested the musculoskeletal model used in the study. MC collected the data and supervised the intervention. MC and DC analysed the data and wrote the first draft of the paper. All authors were involved in the interpretation of the data, in redrafting the manuscript and in approving the final version.

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Ethical approval information

Ethical approval for this study was gained from St Mary's University Ethics Committee.

Written informed consent was obtained from all participants.

Data sharing statement

No unpublished additional data is available from this study.

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$$\begin{pmatrix} \hat{p}_1^1 & \cdots & \hat{p}_M^1 & \hat{p}_{pt}^1 & \hat{q}_1^1 & \cdots & \hat{q}_N^1 & -I_{3\times3} & E_{3\times3} & E_{3\times3} & E_{3\times3} & E_{3\times3} \\ \hat{p}_1^2 & \cdots & \hat{p}_M^2 & \hat{p}_{pt}^2 & \hat{q}_1^2 & \cdots & \hat{q}_N^2 & I_{3\times3} & -I_{3\times3} & -I_{3\times3} & E_{3\times3} & E_{3\times3} \\ \hat{p}_1^3 & \cdots & \hat{p}_M^3 & \hat{p}_{pt}^3 & \hat{q}_1^3 & \cdots & \hat{q}_N^3 & E_{3\times3} & I_{3\times3} & I_{3\times3} & -I_{3\times3} & I_{3\times3} \\ \\ \hat{r}_1^1 \times \hat{p}_1^1 & \cdots & \hat{r}_M^1 \times \hat{p}_M^1 & \hat{r}_{pt}^1 \times \hat{p}_{pt}^1 & \hat{s}_1^1 \times \hat{q}_1^1 & \cdots & \hat{s}_N^1 \times \hat{q}_N^1 & E_{3\times3} & E_{3\times3} & E_{3\times3} & E_{3\times3} & E_{3\times3} \\ \hat{r}_1^2 \times \hat{p}_1^2 & \cdots & \hat{r}_M^2 \times \hat{p}_M^2 & \hat{r}_{pt}^2 \times \hat{p}_{pt}^2 & \hat{s}_1^2 \times \hat{q}_1^2 & \cdots & \hat{s}_N^2 \times \hat{q}_N^2 & \tilde{d}^2 & -\tilde{h}_1^2 & -\tilde{h}_2^2 & E_{3\times3} & E_{3\times3} \\ \hat{r}_1^3 \times \hat{p}_1^3 & \cdots & \hat{r}_M^3 \times \hat{p}_M^3 & \hat{r}_{pt}^3 \times \hat{p}_{pt}^3 & \hat{s}_1^3 \times \hat{q}_1^3 & \cdots & s_N^3 \times \hat{q}_N^3 & E_{3\times3} & \tilde{d}_1^3 & \tilde{d}_2^3 & E_{3\times3} & \tilde{f}^3 \\ \\ \hat{p}_1^{pat} & \cdots & \hat{p}_M^{pat} & \hat{p}_{pt}^{pat} & & E_{3\times N} & & E_{3\times3} & E_{3\times3} & E_{3\times3} & E_{3\times3} & -I_{3\times3} \\ \\ \rho_1 & \cdots & \rho_M & -1 & & E_{1\times N} & & E_{1\times3} & E_{1\times3} & E_{1\times3} & E_{1\times3} & E_{1\times3} \end{pmatrix} \begin{pmatrix} F_1 \\ \vdots \\ F_M \\ \\ F_{pt} \\ \\ L_1 \\ \vdots \\ L_N \\ \\ \hat{R}^1 \\ \hat{R}_1^2 \\ \hat{R}_2^2 \\ \hat{R}^3 \\ \\ \hat{R}^{pat} \end{pmatrix} = \begin{pmatrix} m^1(\hat{a}^1 - \hat{g}) - \hat{S}^0 \\ m^2(\hat{a}^2 - \hat{g}) \\ m^3(\hat{a}^3 - \hat{g}) \\ \\ m^1\hat{c}^1 \times (\hat{a}^1 - \hat{g}) + Y_{3\times3}^1 \ddot{\hat{\varphi}}^1 + \dot{\hat{\varphi}}^1 \times Y_{3\times3}^1 \dot{\hat{\varphi}}^1 - \hat{d}^1 \times \hat{S}^0 - M^0 \\ m^2\hat{c}^2 \times (\hat{a}^2 - \hat{g}) + Y_{3\times3}^2 \ddot{\hat{\varphi}}^2 + \dot{\hat{\varphi}}^2 \times Y_{3\times3}^2 \dot{\hat{\varphi}}^2 \\ m^3\hat{c}^3 \times (\hat{a}^3 - \hat{g}) + Y_{3\times3}^3 \ddot{\hat{\varphi}}^3 + \dot{\hat{\varphi}}^3 \times Y_{3\times3}^3 \dot{\hat{\varphi}}^3 \\ \\ E_{3\times1} \\ \\ 0 \end{pmatrix}$$

... Equation 1

$$\min_{F_i,\ L_j} J = \sum_{i=1}^{M} \left(\frac{F_i}{Fmax_i}\right)^3 + \sum_{j=1}^{N} \left(\frac{L_i}{Lmax_i}\right)^3$$

... Equation 2

where:

$\hat{a}^k$	linear acceleration of the centre of mass of segment *k*
$\hat{c}^k$	vector from centre of rotation of joint at proximal end of segment *k* to centre of mass of segment *k*
$\hat{d}^k$	vector from centre of rotation of joint at proximal end of segment *k* to centre of rotation joint at distal end of segment *k*
$\tilde{d}^k$	skew-symmetric matrix of vector $\tilde{d}^k$
$\tilde{d}_l^3$	skew-symmetric matrix of vector from centre of rotation of hip to tibiofemoral joint contact *l*
$E_{3\times3}$	3×3 matrix of zeros
$\tilde{f}^3$	skew-symmetric matrix of vector from centre of rotation of hip to contact point of patella with the femur
F_i	magnitude of force in muscle *i*
$Fmax_i$	maximum possible force in muscle *i* (upper bound)
$\hat{g}$	acceleration due to gravity
$\tilde{h}_l^2$	skew-symmetric matrix of vector from centre of rotation of knee to tibiofemoral joint contact *l*
i	muscle number
$I_{3\times3}$	3×3 identity matrix
j	ligament number
J	cost function

k	segment number
L_j	magnitude of force in ligament *j*
$Lmax_j$	maximum possible force in ligament *j* (upper bound)
m^k	mass of segment *k*
M	total number of muscles
N	total number of ligaments
$\hat{p}_i^k$	unit vector representing the line of action of force created by muscle *i* that acts on segment *k* (zero if muscle does not insert on segment *k*)
pat	patella
pt	patellar tendon
$\hat{q}_j^k$	unit vector representing the line of action of force created by ligament *j* that acts on segment *k* (zero if ligament does not insert on segment *k*)
$\hat{r}_i^k$	vector from centre of rotation of joint at proximal end of segment *k* to point of action of muscle *i* on segment *k* (zero if muscle does not insert on segment *k*)
$\hat{R}^k$	vector representing *x*, *y* and *z* components of reaction force acting at proximal end of segment *k*

$\hat{R}_l^k$	vector representing x, y and z components of reaction force l acting at proximal end of segment k
$\hat{s}_j^k$	vector from centre of rotation of joint at proximal end of segment k to point of action of ligament j on segment k (zero if ligament does not insert on segment k)
$-\hat{S}^k$	inter-segmental force acting on proximal end of segment k
$-\widehat{W}^k$	inter-segmental moment acting on proximal end of segment k
$Y_{3\times3}^k$	inertia tensor of segment k
ρ_i	ratio of patella to quadriceps tendon forces for muscle i (zero if the muscle is not part of the quadriceps muscle group)
$\dot{\hat{\varphi}}^k$	angular velocity of segment k
$\ddot{\hat{\varphi}}^k$	angular acceleration of segment k